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Development of Patient-Specific Computer Simulation Tools for Endoscopic Scoliosis Surgery

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Abstract

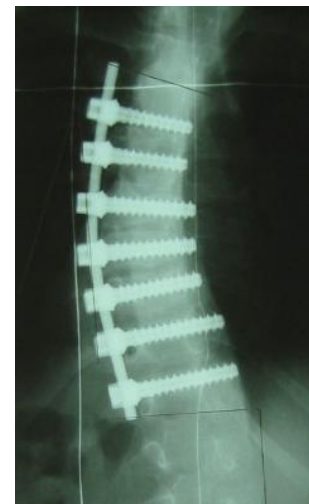
Endoscopic approaches for anterior correction of idiopathic scoliosis are a relatively new surgical technique. This paper describes the development of patient-specific finite element modelling techniques to investigate the biomechanics of single rod anterior scoliosis correction. Spinal geometry is obtained from pre-operative CT scans and material properties for osteo-ligamentous spinal tissues are based on existing literature. The techniques being developed will allow pre-surgical prediction of stresses, forces and Deformations in spinal tissues, rods and screws under post-operative physiological loads.

1. INTRODUCTION

Spine deformity refers to the group of conditions in which the spine exhibits abnormal three-dimensional curvature. The most common type of spine deformity is scoliosis, in which the spine curves laterally when viewed in the coronal plane. The lateral curvature is also accompanied by axial rotation of vertebra and ribcage deformation. The largest single group of spinal deformity patients are children with adolescent idiopathic scoliosis (AIS). Progressive AIS eventually requiring surgical intervention occurs in about two per thousand children, affecting girls much more frequently than boys.

Contemporary surgical interventions for AIS include both anterior and posterior rod systems, in which a single or double rod construct provides curve correction and stability. The relative merits of different instrumentation systems and approaches are still the subject of rigorous debate [1]. One innovative technique for AIS surgery is single rod anterior correction and fusion using an endoscopic (keyhole) approach. Single rod techniques (such as shown in Fig. 1) have shown good radiographic and clinical outcomes provided adequate patient selection is practised [2], and endoscopic techniques for spinal deformity have been under development since 1993 [3]. Such techniques, while relatively recent, appear promising in their ability to correct coronal deformity, restore normal sagittal curvature, reduce scarring and rehabilitation time [4],[5]. As for any spinal instrumentation system, the possibility of biomechanical complications such as screw loosening or rod breakage exists with single rod anterior scoliosis surgery.

This paper presents a methodology for development of patient-specific finite element methods to predict the biomechanical outcomes of scoliosis surgery pre-operatively, with the aim of optimising the performance of instrumentation constructs for anterior single rod AIS surgery. The patient-specific finite element approach is an emerging technology still under development, but has potential to predict surgical correction, compliance of the surgically altered spine, and internal stress/strain in both the osseoligamentous spine and the rod/screw instrumentation system. Fig 1 (Right). Single rod anterior correction for adolescent idiopathic scoliosis



2. METHODS

Development of a finite element model involves prescription of the following parameters: (a) model geometry, (b) material properties for all materials in the model, as well as frictional properties for contacting surfaces, (c) loads and boundary conditions applied to the model, (d) meshing parameters for the model, and (e) numerical solution controls for the model. The following sections describe the prescription of each of these parameters for patient-specific AIS surgery models.

2.1 Model Geometry and Meshing

The geometry for each patient-specific spine model is obtained from pre-operative thoracolumbar CT scans taken in the supine position using a low dose multi-slice imaging protocol. The vertebral column geometry is extracted for the region of interest (ROI) by generating reformatted images through the plane of each endplate in the ROI. Figure 2 below shows a reformatted endplate slice and subsequent finite element representation.

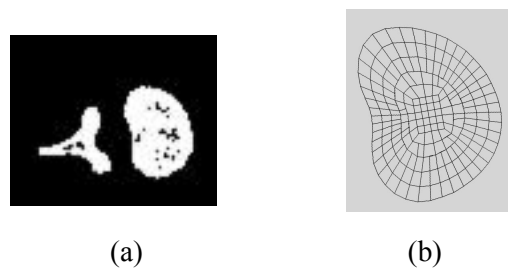


Fig. 2 (a) Reformatted CT scan of endplate and (b) finite element mesh (not to scale)

In the finite element model, the vertebral column cross-section at each endplate is represented by a series of convex elliptical curve segments, together with cubic splines for the posterior concavity. The global position of the geometric centre of each endplate is then measured from the CT stack, and the resulting parameters define the geometric representation of the vertebral column. The intervertebral disc outer profile is implicitly defined from the endplate positions, and the nucleus pulposus boundary may either be defined by prescribing a fixed annular radial thickness, or its position specified directly if sufficient detail is available from CT or MRI images.

The posterior elements (spinous, transverse and articular processes with zygapophysial joint surfaces) are similarly positioned by manual measurement of points from the CT stack. The posterior section of each vertebra is represented using beam elements to reduce solution time for the model, and sliding contact is defined between adjacent zygapophysial joint surfaces as shown in Figure 3(a). The orientation of zygapophysial joints is specified using card angles as defined by Panjabi et al [6]. A custom-written finite element pre-processor generates the finite element mesh for the entire model based on the geometry measurements just mentioned and according to user-specified meshing parameters. A finer mesh provides greater solution accuracy but increased solution time, and vice versa. The pre-processor also performs automatic attachment of ligament connector elements to the posterior processes, assignment of articulating contact surfaces, material properties, and loading cases, providing rapid model generation once the geometric parameters have been extracted from each CT dataset.

2.2 Material Properties

Material properties for the scoliotic spine model are currently based on published values for cortical bone, cancellous bone, intervertebral disc ground matrix, collagen fibres, articular cartilage, ligamentum flavum, interspinous and supra-spinous ligaments, and intertransverse ligament material behaviour. Material properties for biological tissue can vary widely from patient to patient, and one of our current focus areas is patient-specific approximation of soft tissue material properties based on in-vivo tests such as axial tension/compression and ligamentous laxity tests.

2.3 Loads and Boundary Conditions

Using the supine position as the ‘zero load’ reference, the finite element model is subjected to loading about axes corresponding to flexion/extension, left/right lateral bending, and trunk rotation to simulate real patient movements. Since the model does not include muscles, this loading is applied at the superior vertebrae of the model, and resisted by an encastre boundary condition at the inferior vertebrae. This loading regime simulates the usual method of load application for mechanical in-vitro spine tests.

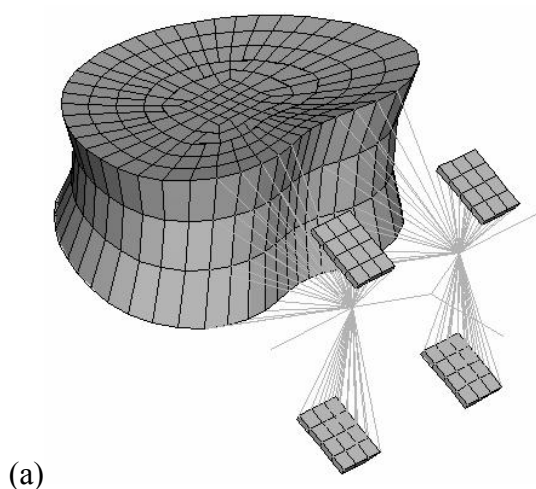
2.4 Numerical Solution Controls

The finite element model is solved using a quasi-static solution (inertia effects neglected) using the ABAQUS/Standard software. Default criteria for solution convergence and time incrementation are used.

3. SIMULATING SURGICAL INTERVENTION

As mentioned, the aim of the finite element methodology is to predict biomechanical outcomes of surgery, and this is achieved by comparing the predicted loading response of the intact (but scoliotic) pre-operative spine, with the loading response of the post-operative instrumented spine, before any bony fusion has occurred between vertebrae. The finite element methodology is thus used as a comparative tool, and uncertainties regarding patient specific material properties are therefore less critical since the emphasis is on predicting differences between the pre-operative and post-operative spine.

Figure 3(b) shows a coronal CT reconstruction for an AIS patient. This patient’s major curve extended from T5 to T11, so the finite element model was generated over this region as shown in Figure 3(c). The surgically altered model is shown in Figure 3(d) after discectomy and screw placement, and prior to rod insertion. The total time required for generation of the finite element model for this patient was approximately six hours, with manual measurement of spine geometry from the CT stack accounting for most of this time. Actual solution time for each finite element model using ABAQUS/Standard is expected to be around four hours, making patient-specific pre-operative planning for scoliosis surgery a feasible option at least in terms of processing time per patient.



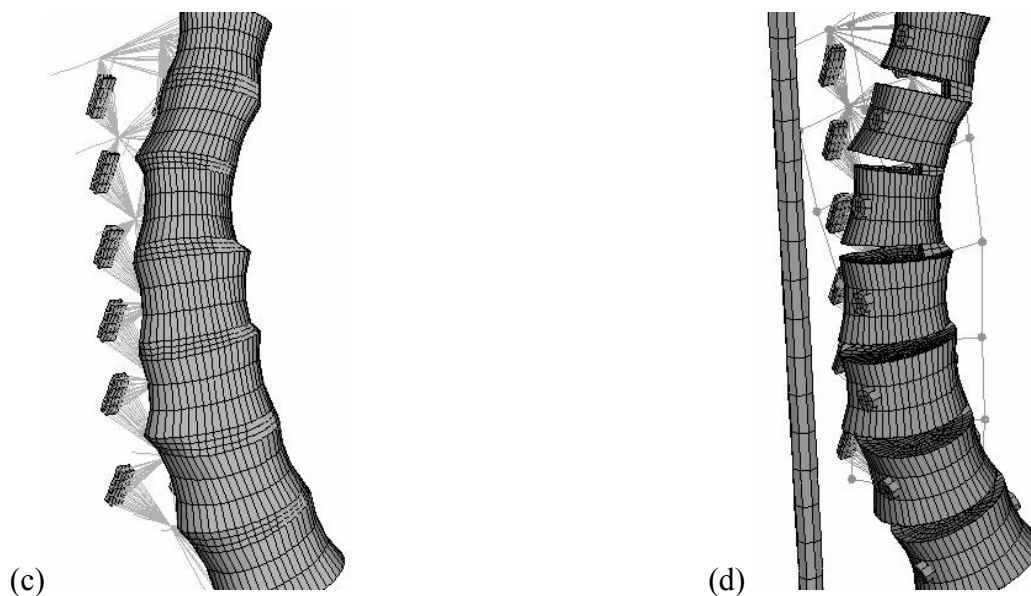


Fig. 3 (a) Finite element mesh of a single vertebra, (b) Coronal CT reconstruction, (c) Pre-operative finite element model, (d) Intra-operative finite element model with discectomy, anterior rod, and vertebral body screws

4. CONCLUSION

A finite element methodology has been developed for patient-specific simulation of single anterior rod surgery for adolescent idiopathic scoliosis. Issues to be addressed in further development of the model include the prescription of patient-specific material properties, analysis of inter and intra-observer errors associated with geometry measurement from CT scans, and validation of the methodology by comparison of predicted and actual biomechanical outcomes for AIS surgery patients.

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